

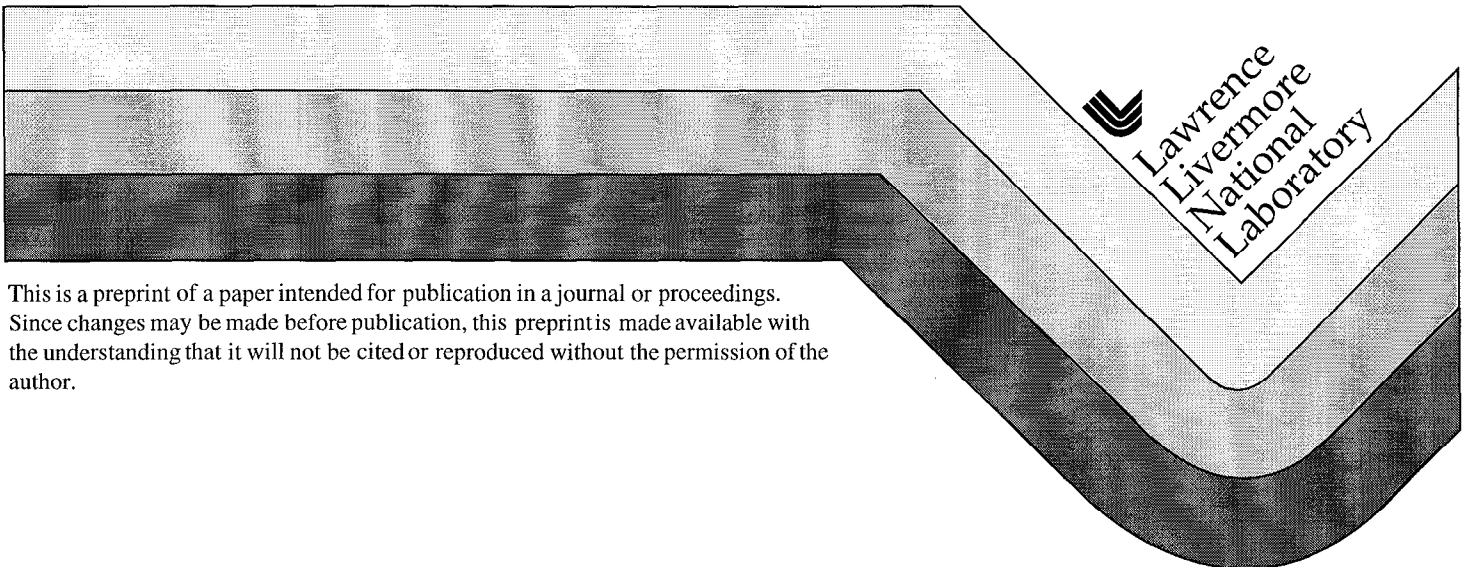
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Medical Applications of Ultrashort Pulse Lasers

B.-M. Kim
M. D. Feit
A. M. Rubenchik
L. B. Da Silva

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B.-M. Kim (kim12@llnl.gov), M. D. Feit (feit1@llnl.gov),
A. M. Rubenchik (rubenchik1@llnl.gov), L. B. Da Silva (dasilva1@llnl.gov)

Medical Technology Program, Lawrence Livermore National Laboratory
7000 East Ave. L-399, Livermore, CA 94550, USA
Phone : 925-423-3262
FAX : 925-424-2778

Abstract

The characteristics of the ultrashort pulse laser (USPL, < 1 ps) ablation of biological tissues are investigated both theoretically and experimentally. Effective USPL parameters for minimal damage and high ablation rates are discussed.

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Introduction

USPL ablation of matters shows unique characteristics that are distinctive from longer pulse ablation¹⁻⁵. Since the pulse width is extremely short, the beam intensity is high even with moderate energy per pulse. When this intense pulse interacts with matters and the beam intensity exceeds the threshold, optical breakdown occurs and a plasma is produced. Electrons in the plasma then efficiently absorb laser energy and through collisions heat up and further ionize the plasma. The plasma is then expanded carrying off most of the deposited energy. In the case of USPL, this procedure takes place very quickly and little energy is transferred to the lattice. Therefore, USPL ablation causes minimal thermal damage to the surrounding material. On the other hand, for long pulses, the energy is transferred to the lattice significantly and the material is heated in a large volume inducing extensive collateral damage.

Results and Discussion

The USPL ablation procedure was studied using computer simulations. Even if the initial pressure and temperature created during optical breakdown was so intense (~ 1 Mbar and ~ 20 eV during water ablation), it was clearly shown that the fast ejected plasma carries most of the energy to the free space leaving little effect on the surrounding materials. This finding was confirmed by the morphologic studies of USPL and long pulse ablated craters as shown in Figs. 1 and 2. The ablation craters created on soft and hard tissues using USPL show minimal thermal or mechanical damage to the surrounding region indicating extremely accurate and precise tissue removal is possible.

There has been much effort to use this sophisticated technique for medical applications where precise tissue removal and minimal collateral damage is needed. However, further experimental and theoretical studies revealed that when non-optimal laser parameters are used, thermal damage can be caused to the material even with pulses as short as 100 fs. Also, it is well known that it is harder and much more expensive to build femto-second lasers than to build pico- or nano-second lasers. It is then questionable if it is necessary to use femto-second lasers rather than pico-second lasers. Experimentally we have determined the parameters for the "optimal" laser that would achieve efficient ablation with minimal collateral damage. First of all, experimental and theoretical studies indicate that the longest pulse width that creates the desired high quality ablation craters was approximately 1 ps. Studies on ablation threshold, crater morphology, and pressure generation all indicate that there might be a transition in ablation quality for pulse widths between 1 ps and 5 ps. Secondly, it was found that the ablation rate ($\mu\text{m}/\text{pulse}$) was similar for the first 100 shots (when using 1 kHz rep rate and 130 fs pulses) independent of the beam intensity. The crater depths were measured using 2x, 4x, and 7x ablation threshold intensities and the ablation rate was effectively constant at 1 $\mu\text{m}/\text{pulse}$. This result indicates that intensities above threshold do little to increase ablation but instead increase the amount of energy deposited into tissue and therefore increase collateral thermal/mechanical damage. Finally, the parametric studies on hard tissue surface damage revealed that even with 100 fs pulses, thermal damage can occur for large beams when the repetition rate is high. We also observed that the surface damage threshold decreased by 30% at repetition rates of 1 kHz as the beam diameter increases from 130 μm to 260 μm . The threshold did not change significantly with beam size when the repetition rate is 100 Hz or lower. Since the optical breakdown threshold should not depend on the repetition rate and beam size, it was concluded that the observed change in ablation threshold was induced by thermal accumulation. Therefore, when the intensity is less than the optical breakdown threshold but higher than the thermal damage threshold which is highly dependent on repetition rate and beam size, significant thermal damage is possible. It should be noticed

that the thermal damage due to high repetition rate and large beam size takes place only when the beam is focused on a single spot for a long time. Since the beam size used for USPL ablation is generally quite small (100s of microns), it is unlikely to focus the beam on the same spot for a long time and that this would be hardly a clinical issue. However, since even higher repetition rate might be needed to improve the ablation rate for broader clinical usage, it is desirable to have a beam scanning module to eliminate possible thermal damage.

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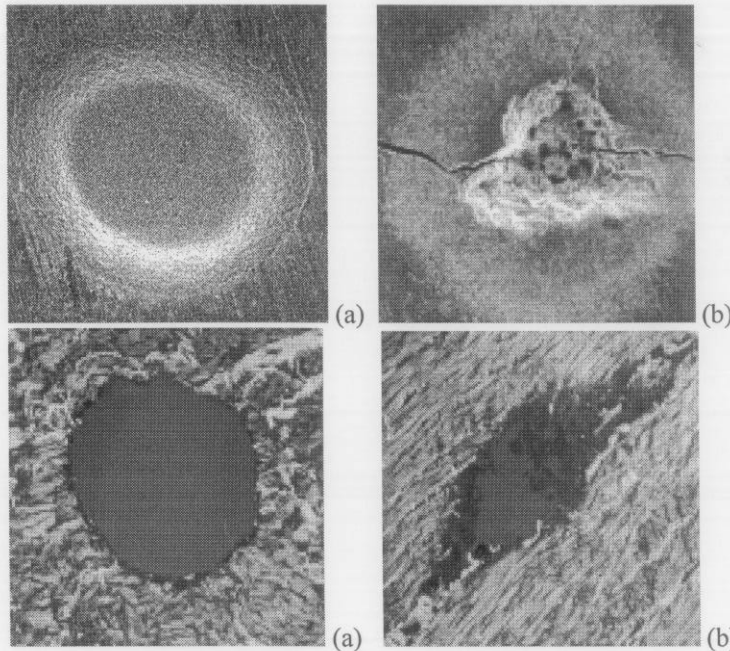


Fig. 1. Ablation craters created on hard tissue (tooth) using (a) USPL (pulse width = 500 fs) and (b) long pulse laser. (pulse width = 1 ns). Mechanical cracks are observed in the long pulse crater.

Fig. 2. Ablation craters created on porcine myocardium using (a) USPL (pulse width = 500 fs) and (b) long pulse laser. (pulse width = 1 ns). The birefringence loss of heart muscle fibers due to thermal damage is observed around the crater when long pulse was used.

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